Design of a Sensor Based Data Collection System for Lower Limb Prosthetic Gait Analysis

by:

Thomas Campbell Bulea

An undergraduate honors thesis submitted in partial fulfillment of
requirements for graduation with distinction in
Mechanical Engineering
The Ohio State University
2005

Honors Committee:
Advisor: Dr. Eric R. Westervelt
Honors Chair: Dr. Mark Walter

ABSTRACT

Design of a Sensor Based Data Collection System for Lower Limb Prosthetic Gait Analysis

by:

Thomas Campbell Bulea

Advisor: Dr. Eric R. Westervelt

The primary function of a lower limb prosthetic device is restoration of ambulation. Proper alignment – the correct spatial relationship between artificial sockets and the natural limb – is paramount to attain an efficient, comfortable gait with a desired loading pattern on the residual leg. Despite advances in prosthetic device design, the clinical alignment process remains subjective and nonsystematic due to a lack of an inexpensive, effective method for quantification of the amputee gait. Gait laboratories provide accurate data for gait monitoring; however cost and lab availability prohibit most patients from this benefit. Economic concerns aside, gait labs do not fill the void of information needed to quantify the alignment process. Observation time and environment are too limited to amass useful information for prosthetic alignment improvement. A more logical and systematic approach to clinical alignment requires the quantification of amputee gait before and after adjustments made by the prosthetist. To be complete this quantification must span extended periods of time and terrain. Thus, there is a patent need for a portable, reliable, and cost effective motion capture system.

This project proposes a design for such a system. Comprised of body (prosthetic) mounted inertial sensors – accelerometers and gyroscopes – the system is designed to track the kinematics of the limbs during a walking cycle. The goal of this work is to prove the feasibility of motion capture system using these body mounted sensors. The effectiveness of the system will be judged as its ability to capture planar motion using two (2) accelerometers and one (1) gyroscope mounted on an aluminum bar (simulating a prosthetic device). The design of the system was formulated based on an extensive literature review pertaining to body mounted sensor systems. The rigid structure of the prostheses gives a prosthetic mounted sensor system a distinct advantage over a body mounted system in terms of inverse kinematic calculations.

For my father, Craig T. Bulea.

ACKNOWLEDGEMENTS

First and foremost I would like to thank my undergraduate advisor, Dr. Eric R. Westervelt, co-director of the Locomotion and Biomechanics Lab. The most motivated and dedicated professor I have ever been fortunate enough to work with; this project would not have existed were it not for his effort and thirst for knowledge. I would also like to thank the second co-director of the LAB Lab, Dr. James Schmiedeler, and all other members for their support of this work.

Lastly, I would like to thank my family and friends for their never ending support during my undergraduate career at Ohio State University. Special gratitude is due to my parents: Craig and Nancy Bulea, brothers: Craig, Doni, and Tony, sister: Erin, and of course to Nikki. The strength you have given me is invaluable.

TABLE OF CONTENTS

LIST OF FIGURES

CHAPTER 1

Introduction

1.1 Project Motivation

The main function of lower-limb prostheses is restoration of ambulation (walking or running). Advances in the fields of materials science and engineering in the past several decades have greatly enhanced the ability of prosthetic devices to restore this function to the patient. Proper alignment - the correct spatial relationship between the prosthesis socket and residual limb - is paramount to enabling an efficient, comfortable gait. Correct alignment ensures a natural gait and a desirable loading pattern on the residual limb [1,2]. Lower limb prosthetic devices are initially aligned during manufacture using a process called bench alignment. The bench alignment process involves the adjustment of all replacement joints on the device until they “appear” to align in a proper position to accomplish a ‘natural’ gait. Typically, a mechanical jig, similar to the one below in Fig. 1, designed specifically for each type of prosthetic (i.e. above-knee, below-knee, etc.) is used during this bench alignment process. This jig provides guidance; however it does not provide a totally systematic process.

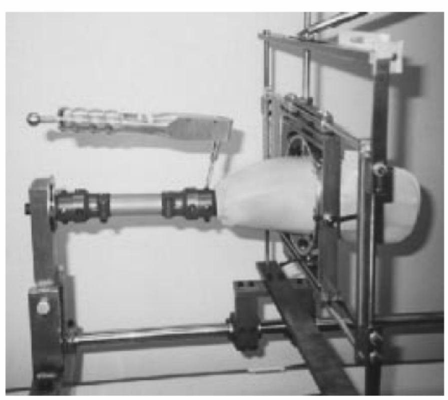

Fig. 1: An example of a mechanical jig used for bench alignment of lower limb prostheses. Taken from [3].

Clinically, alignment is obtained through a dynamic procedure using subjective assessments of the gait pattern by the prosthetist as well as subjective feedback from the patient. Inherent to this dynamic alignment phase is the problem with the clinical method; the quality of the alignment is based upon heuristics. In a study performed by Zahedi, et. al it was demonstrated that the definitive alignment achieved for a transtibial amputee using these subjective assessments was never unique, but instead actually fell in a large range [4]. In 2001, Sin, et. al. conducted a detailed study of clinical alignment in an attempt to quantify the wide range of effective positions [3]. Prior to their analysis, the group determined the effects of alignment in the sagittal plane to be the most consequential during clinical alignment; hence their average acceptable range was quantified. These sagittal plane alignments consist of two adjustments: anterior-poster (A-P) shift (mm) and A-P tilt (degrees). The maximum acceptable alignment positions for level and non-level walking was quantified for six patients. In all cases, the acceptable alignment range for non-level walking fell within that for level walking [3]. The boundaries for acceptable alignment in each case (level or non-level) were different for each patient; however, the averages of acceptable levels were determined and a critical alignment zone was identified (see Fig. 2). Alignments within this zone were acceptable to all test subjects on both level and non-level walking surfaces. The study supports the concept of a wide range of satisfactory alignment positions for each patient. These satisfactory alignments may be functional initially, but in time undue may be imposed on the residual limb unless the definitive alignment position is reached.

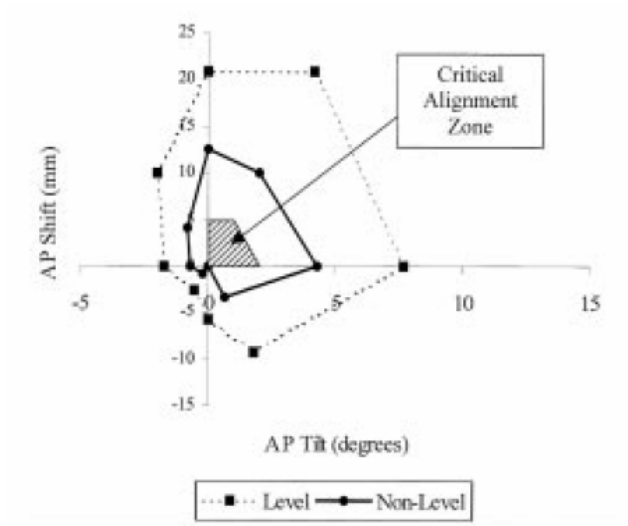

Fig. 2: Averaged acceptable alignment zones for level and non-level walking and the determined critical alignment zone. The overall range for A-P shift was -15 to 35 mm for level walking and -10 to 20 for non-level walking. The range for A-P tilt was -5 to 13 degrees for level and -4 to 5 for non-level. Taken from [3].

Often times, this undue stress on the residual limb will cause new prosthetic device users to return after the initial prosthesis alignment for further adjustment. Because there is no method to quantify the results of the alignment adjustments, the patient is continually subjected to a satisfactory alignment rather than a definitive one. If these return trips are still ineffective the device will remain cumbersome, painful, and unreliable for stable walking; deterrents that may cause the patient to abandon use of the device.

Although it is rarely done, due to cost restrictions and lab availability, a standard gait analysis lab may be used to enable more accurate identification of the patient's gait [5]. In these labs, sophisticated sensors and cameras are attached to the patient. The patient is then observed as he passed through several walking or running cycles. The objective of such analysis is to collect enough data to accurately quantify the patient's gait during use of the device. In this way, the prosthetist can observe quantitatively any abnormalities present in the gait of the patient (e.g.

heel whip or over flexion of the knee). While this method seems to offer obvious advantages over the "feel" technique used by most doctors, it does have drawbacks. First, the physical environment of a gait laboratory is not conducive to accurately represent the environment in which the patient will utilize the prosthesis; gait labs are typically too uniform for an accurate re-creation of the patient's everyday use. Second, the time of observation is much too small. Patients are not inanimate and as such, they will continually interact and adapt to their prosthetic device [6]; these interactions must be observed to obtain an accurate assessment of the device's functionality.

Long term observation issues, cost restrictions, and availability render gait labs ineffective for clinical use in alignment of a prosthetic device. Moreover, the fact that gait labs are able to accurately capture artificial limb motion is irrelevant to the alignment issues because the data does not quantify the effects of re-alignment. The gait lab may be able quantify problems with the gait, but it is still the responsibility of the prosthetist to make the necessary adjustments to the device to correct the mistakes. Then, once the remedy is complete the prosthetist has only limited time to observe his patient and verify his actions were the correct ones.

A more logical and systematic approach to prosthetic device alignment must involve the quantification of the gait before and after the adjustments made by the prosthetist. Furthermore, to be complete this quantification must be made over extended periods of time and terrain. Thus, there is a patent need for a measurement system with the ability to track the motion of a prosthetic device during waling cycles before, during, and after alignment. To accomplish this, a more efficient means for observation is necessary.

1.2 Project Description

The project outlined in this thesis is rooted in the above discussion. The project's contribution is in the development of a wearable motion capture system that is portable and effective in its results. Advances in sensor technology in the past decades have made possible the concept of a wearable measurement system consisting of multiple sensors and a data logging system capable of accurate motion capture. There have been many attempts at the creation of motion capture systems that make use of this technology; section 2 below contains a comprehensive review of such devices. The majority of these systems have been applied directly to humans and not to prosthetic device design. Uncontrollable errors are introduced for a variety of reasons when sensors are attached directly to human subjects, including, but not limited to sensor placement repeatability error, movement between externally attached sensors and the skin, and relative subcutaneous movement between the musculature and dermis.

Motion capture of a prosthetic limb's motion while walking has many advantages over the motion capture of a healthy limb due to the nature of the device. A prosthetic device is a rigid mechanism thus; sensors can be firmly mounted - via screws or other mounting methods- securely to it. This eliminates many of the errors introduced by attachment of sensors to the human body and skin. The rigid nature of the device also simplifies the mathematical analysis needed to capture motion from the sensor signals and provides a known reference frame with a known center of mass.

1.3 Project Objectives and Contributions

The brief discussion of the previous paragraph outlines the basic approach followed in this project. The project involved the design and fabrication of a prototype measurement system consisting of micro-electro-mechanical sensors (MEMS) mounted on a rigid bar. The placement and configuration of the sensors on a bar, which represented a lower-limb prosthetic device, was used to allow study of the effect of sensor placement. Features of the developed system include the ability to track planar motion using only three (3) small, lightweight sensors at a cost of less than $200 for the sensing equipment. This cost does not include data acquisition hardware. An algorithm was developed for analysis of collected data with the purpose of reconstructing the kinematics of the gait cycle (movement of the prototype system). The system itself has been briefly tested for feasibility in low frequency motion capture similar to that of a walking cycle. Other applications for such a device include the estimation of planar biped motion – a field closely tied with prosthetics.

CHAPTER 2
Background Literature Review

2.1 Traditional Motion Analysis Systems

The study of ambulation has been undertaken by many researcher groups for more than one hundred years. One of the first known "gait monitoring" systems was Marey's [7] walking device applied in 1874, seen in Fig. 3. While this antiquated system may seem far removed from today's technology, the concepts of monitoring acceleration and position of the limbs during gait cycles remain central today.

Fig. 3: Marey's runner outfitted with clockwork recorder and accelerometer. Originally published by Marey 1874, taken from [7].

As previously discussed, one product of modern technology that has allowed researchers to accurately monitor limb motion is the gait laboratory. .Gait labs typically employ some type of vision tracking system – camera or infrared – to track specific points on the subject as they ambulate. A typical gait laboratory is setup as a relatively long walkway with several sensing components (see Fig. 4). Perhaps the most important parts of the system are the markers placed on the subject and the detectors used to track them. These markers can be either passive or active, as can the detection systems. In addition to the tracking system, the floor of such a setup usually contains force plates which measure ground impact reactions during the walking cycle.

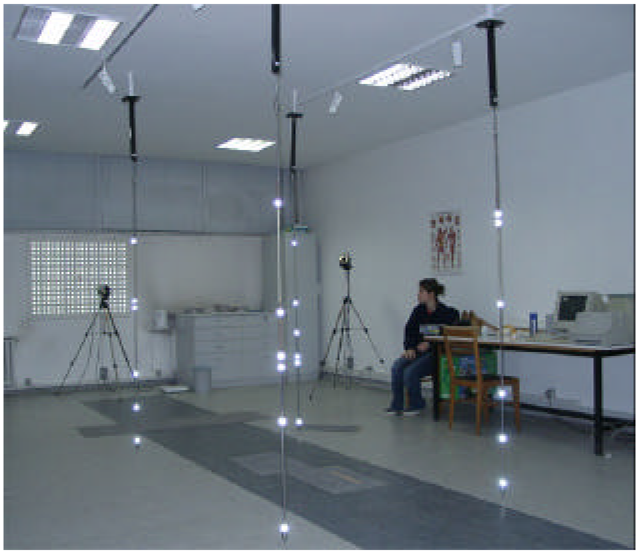

Fig. 4: Photograph of a gait laboratory with a vision system using passive markers. Taken from [8].

While these types of gait laboratories are accepted as accurate and repeatable [9]; they do have drawbacks. As mentioned above, duration and terrain are of concern when gait is being monitored in a controlled lab setting, which does not accurately portray the prosthetic's everyday use by the patient. In addition to these limitations, data collection procedures are long and complicated. Each patient's test can take up to two (2) hours and requires an engineer (or technician) and clinician to be on hand. The data collected can be processed within minutes to yield numerous desired parameters for each patient's gait, yet it takes weeks to receive a clinical report [9]. Most reports given to the clinician are the same regardless of the desired information; thus it is often too complicated to benefit the clinician in the end. The whole procedure is expensive and inefficient; the average cost of one patient session in 2000 was roughly $2000 Canadian [10]. In the U.S. it is rare to find a gait lab where revenue balances the expenses [9].

The benefits of gait analysis are well known, yet the cost of using gait lab analysis as a clinical tool is prohibitive. However, the use of MEMS sensors for cheap, portable gait measurement systems is promising. A concept for such a system is discussed next.

2.2 Measurement Systems with Electronic Sensors

As mentioned, continuously advancing circuit technology has led to the invention of scores of sensors for applications in fields ranging from robotics to aerospace to biomedical to industrial. These sensors are becoming more compact and less expensive every year [11]. A brief review of literature reveals the use of many small transducers and sensors in the arena of gait analysis including such devices as: electrogoniometers, gyroscopes, inclinometers, accelerometers, and force sensitive resistors. These devices have potential to be applied for measurement of such quantities as joint angle, limb angular velocity, limb tilt angle, linear acceleration, foot impact forces, and foot contact time; more than enough parameters to perform useful gait analysis on a patient. In fact, several systems have been designed to use these inertial sensors to monitor gait.

One of the seminal groups to attempt a gait analysis system using small, modern sensors was that of Dr. A.J. van den Bogert and his research group from the Human Performance Laboratory in Calgary, Canada. The group proposed a measurement system, seen in Fig. 5 below, containing four (4) tri-axial accelerometers mounted on the upper body of the patient [12].

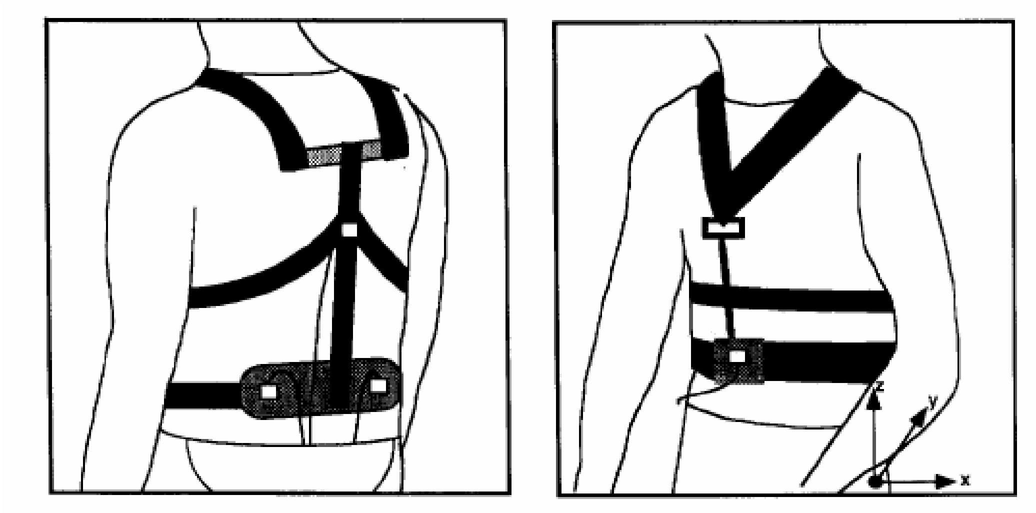

Fig. 5: Front and back views of van den Bogert's portable body segment measuring system. Three accelerometers are mounted on the back of the patient and one is mounted on the front. The signals are recorded with a small data logger carried in the belt. Taken from [12].

The goal of their system was to accurately capture total resultant force and moment on a body segment, in three dimensions, from accelerometer data. The group used inverse dynamics to derive the inter-segmental loading pattern on the hip during the single support phase of the walking cycle. The measurement system was evaluated versus standard gait laboratory analysis; the results indicated that the portable measurement system underestimated force and moment at the hip by approximately 20% [12]. The reason for this shortcoming was most likely rooted in assumptions made during the inverse dynamic analysis. Namely, the body and leg were assumed to be a rigid body; this is probably not the case during the beginning (impact) and end (toe off) of the stance phase. Also, the inertial and gravitational effects of the swing leg on the acceleration of the torso were neglected; this too could be a source for error. Despite these shortcomings, the system does have advantages over traditional gait lab systems, especially if the underestimations are acceptable for the particular application. The primary advantage is the system does not require a lab setting, allowing for long term field study under many environments and terrains.

Also, the inverse dynamics used did not include integration or differentiation; hence real time gait assessment may be possible with the portable data logging system.

While van den Bogert's system provided a method for portable assessment of gait, it also showed some weaknesses of body mounted sensor systems in tracking gait dynamics. Because of these difficulties, research began to focus on a method proven to be successful by gait laboratories, that is, tracking the kinematics of body segments. Thus, systems were invented which mounted small electronic sensors directly on the body segments for purpose of tracking position, velocity, and acceleration. One of the most successful of these systems was introduced in 1999 by Tong and Granat [13].

Tong and Granat developed a portable gait monitoring system using only uni-axial gyroscopes mounted on the body [13]. A gyroscope was attached over the skin on both the thigh and shank using a Velcro strap; see Fig. 6. The angular velocity (in the sagittal plane) was recorded for each segment. These two signals were then used to derive the inclinations and knee angle of each segment. These measurements and calculations were then compared with data taken from a Vicon vision system in a gait laboratory. The correlation between segment angular velocity, knee angle, and segment inclination was excellent for straight walking, with less than 10% difference between the two systems [13].

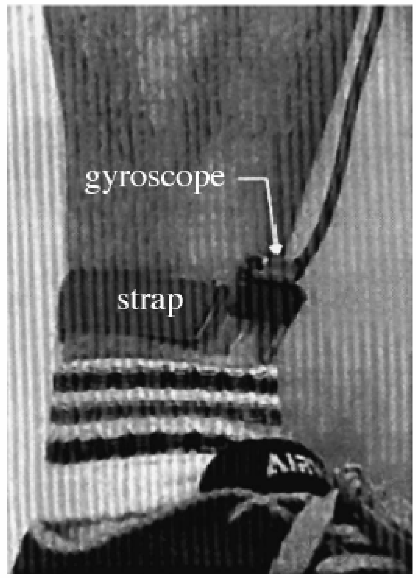

Fig. 6: Tong's strap-gyroscope system for portable gait analysis. Only the shank segment is picture, but a gyroscope was attached to the thigh in a similar manner. Taken from [13].

The majority of these errors were believed to be due to subtle movements of the sensor during the walking cycle. These movements are most likely a result of compliance in the attachment method of the sensor. Problems with correlation of the integrated data – inclination and knee angle - arose when the subjects turned while walking; the signals appeared to drift from the Vicon system data after integration. Tong and Granat proposed a solution to this drift error based on the visual observation of an initial inclination of each segment during the turning phase. Their "automatic reset" technique seemed to quell the integration drift errors introduced during this turning phase. This reset technique assumes the thigh and shank segments are vertical during the mid stance phase, meaning the inclination for both is zero at that time. A force sensitive resistor (FSR) was placed under the heel, and the mid stance phase was indicated by the peak signal from this sensor. At the moment the peak signal from the FSR was seen, the inclination of the thigh and shank were both reset to zero. In addition to showing the feasibility of a portable system to track kinematics of body segments based on gyroscope signals, this group also proposed methods for the derivation of the number of steps, walking speed, and stride length from those same signals.

Another group that compared a body mounted sensor system with a Vicon gait laboratory system was led by Anand Nene of the Roessingh Rehabilitation Center (Enschede, The Netherlands). His group compared measurements taken from biaxial accelerometers with those from the Vicon vision system [14]. The position of the accelerometer placement on the thigh and shank can be seen below in Fig. 7.

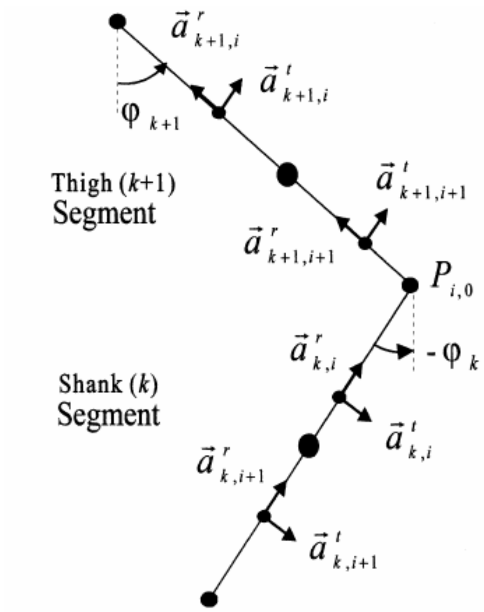

Fig. 7: Placement of Nene's biaxial accelerometers on both the thigh and shank segments. Taken from [14].

The two signals (tangential and radial) from each accelerometer were used to find the linear and angular acceleration of segment, which in turn was used to compute the moment about the knee joint. In addition, the angular velocity of the joint was determined by the integration of the accelerometer signals. This was compared with signals taken from a gyroscope and a Vicon vision system. The accelerometer signals were then integrated again to solve for the absolute angles of both the thigh and the shank. The results of the accelerometer system were excellent when compared with the data taken from the gait laboratory. Part of the increased success of this sensor system lies in the mounting method used; Nene and his group placed accelerometers on a metal strip in order to create a more rigid structure and eliminate error introduced by relative movement between mounted sensors [14].

The previous systems described indicate the feasibility of a reliable, accurate, and portable gait measurement system consisting of small, body mounted sensing networks. Despite their obvious potential, all systems have limitations. These include problems with drift in data, movement of sensors during data collection, and errors introduced from the assumption of rigid behavior of human limbs. Moreover, systems can become cumbersome difficult to use if a multitude of sensors are required for accurate motion capture.

Thus, a practical system must be small, easy to apply, and provide the desired information accurately and reliably. The design of such a system is undertaken in the next section.

CHAPTER 3
Measurement System Design

3.1 Initial Design Considerations and Assumptions

The cursory review of several research groups' work above revealed numerous attributes and shortcomings of motion capture systems made from body mounted sensors. Without a doubt these systems enjoy numerous advantages over gait laboratory systems, the decisive one being their extreme mobility. Body mounted sensor systems are far more cost effective than gait labs since they do not require constant presence of a clinician or engineer. They also provide more direct and efficient data capture over wider variety of terrain for a longer period of time. The data collected gives direct information about the kinematics of limbs, which could potentially be more useful to clinicians than a report given by a gait lab system, indicating the possibility for more effective clinical performance. Finally, these systems are more convenient for the patient to use than participation in a gait lab study. The background survey also disclosed multiple problems with body mounted sensor measurement systems. The chief concern of most systems is data integrity regarding accuracy and repeatability. The data from sensors is often processed heavily, revealing intrinsic errors in many techniques (e.g., data integration). Body attachment method can be responsible for lack of accuracy and repeatability. Concern also arises when multiple sensors are used, creating a cumbersome feel for the patient, which may affect gait pattern and results. The challenge for the designer is then to accentuate the many benefits of the body mounted sensor system while attenuating the shortcomings.

The scope of this system lies in the arena of prosthetic devices, whose constraints provide beneficial – and simplifying – effects. Prosthetic devices, unlike natural limbs, are made from materials which behave as rigid structures [15]. This rigid behavior of the links provides the opportunity for vastly improved inverse kinematic calculations, and elimination of errors produced from sensors being mounted to the human skin. For conspicuous reasons, it would not be advantageous to permanently mount (via a screw or plate) sensors onto the limbs of a human patient; however, prosthetic devices are assembled from interchangeable components, some of which could be outfitted with sensors rigidly attached. These rigid attachments would alleviate any relative movement between the sensors and the limbs, a major source of body mounted system error [14]. Since the mounted sensors effectively become a part of the amputee's prosthetic device, the testing process will be much less cumbersome than it would for a person with natural limbs. This increased tolerance allows for the inclusion of more sensors in the system without significant impedance to their gait cycle.

Finally, it is important not to lose sight of the purpose of this design, to help aid and quantify the dynamic alignment process for lower limb amputees. Discussed at length in the introduction, the importance of the dynamic alignment process can not be overemphasized. The clinical process of dynamic alignment is based on subjective assessment by the prosthetist and is sometimes influenced by the opinions of the patient. The clinical practice of dynamic alignment consists of two manual adjustments to the device; A-P shift and A-P tilt [3]. The term A-P refers to linear and angular adjustments made in the sagittal plane (see Fig. 8 below). Thus, the goal of the designed system is a proof of feasibility with regards to quantifying the results of these subjective adjustments using mounted electric sensors.

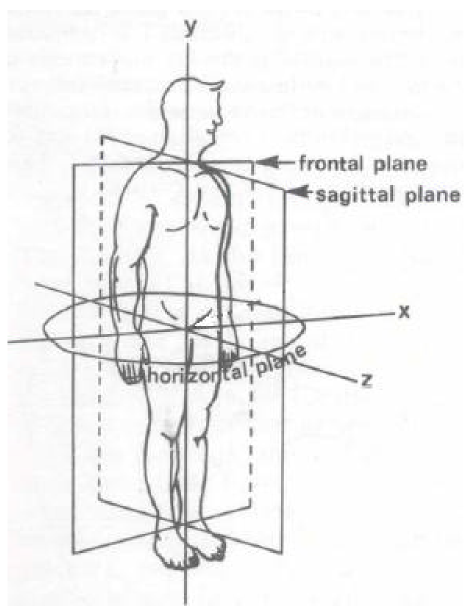

Fig. 8: The human planes of reference. The sagittal plane divides the body into left and right halves. The frontal plane is perpendicular to the sagittal plane and divides the body into front and back sections. The horizontal or transverse plane divides the body into top and bottom halves. Taken from [16].

The complete quantification of the results would require motion capture in all three of the planes of reference. Since the aim of this system is to prove the feasibility of quantification via mounted sensors, the prototype system was designed to capture motion in only one plane. The representative plane was chosen as the sagittal for two reasons: (1) adjustments made during dynamic alignment are performed here and (2) the effect of gravity is present. The term representative will serve to indicate the assumption that feasibility in the sagittal plane will correlate to feasibility in all planes.

3.2 Choice of Electrical Sensors

Using the information gained from the background survey, two sensors appeared to show the best results for motion capture during walking cycles: accelerometers and gyroscopes [12-14].

3.2.1 Accelerometers

A single axis accelerometer consists of a very small mass connected to a spring within a housing to provide resistance against displacement (see Fig. 9) [17]. This displacement is proportional to the difference of acceleration and gravity along the sensitive direction.

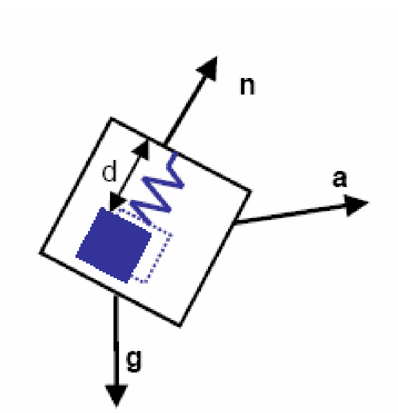

Fig. 9: Schematic of a single axis accelerometer. Displacement **d** is a function of the acceleration **a** and gravity **g**. Taken from [17].

A multi-axial accelerometer can be constructed in two ways. The obvious method is to employ multiple mass and spring housings to obtain measurements in different directions. The second way is to employ one mass with multiple translational degrees of freedom – one for each sensitive direction. Over the past decade, the use of single mass, multi-directional accelerometers has allowed for fabrication of extremely small sensors which can be worn on the body. These micro-electro-mechanical sensors (MEMS) employ extremely small masses suspended by polysilicon springs in micro-machined silicon housings mounted on silicon wafers. Deflection of the small mass is measured and converted to a voltage signal via a differential capacitor [18].

3.2.2 Selection of Appropriate Accelerometer

The desired accelerometer had to meet several basic requirements. First, it must have a dual-axis output in order to monitor acceleration in both directions of the sagittal plane. Size constraints required the sensor to be mounted on a plate no larger than one (1) square inch. Because cost was a limiting factor in the project, the sensor was to cost under $100. In addition to these fundamental requirements, several application specific criteria had to be considered and are discussed in the following paragraph.

In low gravity applications like motion capture, the dominating constraint for choice of a MEMS accelerometer is its limiting resolution, i.e. its minimum detectable input value [19]. The limiting resolution is directly proportional to the measurement noise floor of the instrument, which is controlled by the bandwidth of the measurement being taken. Review of the literature revealed walking cycle signal frequencies to be at or below 5 Hz [13, 14]. Thus, the ideal accelerometer will be adjustable to small bandwidths, yielding a high signal to noise ratio. The second application specific constraint requiring consideration for sensor selection was the magnitude of accelerations created by walking motion. Consultation of the literature showed a maximum acceleration at any point on the natural limb of between 5 and 10 *g*'s [16].

After consideration of the above criteria, the ADXL210 accelerometer from Analog Devices was chosen. The ADXL210 is a +/- 10 *g*, dual axis accelerometer with a pulse width modulated (PWM) digital output. The sensor itself is contained on a monolithic integrated circuit, which converts the output circuitry to convert the analog acceleration signal to a PWM output signal. The sensor was purchased in combination with an evaluation board (ADXL210EB). This board provided circuitry to provide convenient adjustment of the sensor bandwidth and duty cycle period. A schematic of the ADXL210 can be seen in Fig. 10.

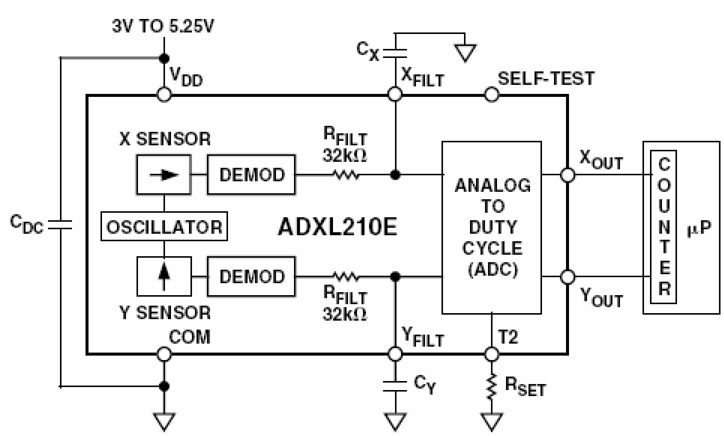

Fig. 10: Functional Block Diagram of the ADXL210. PWM output period is controlled by R_{set}. Bandwidth and noise floor are controlled by C_x and C_y. Taken from [20].

3.2.3 Gyroscopes

The basic operating element of a MEMS rate gyroscope is a vibrating piezoelectric element coupled to a sensing element. The input voltage to the sensor is used to drive the piezoelectric element to resonance. The velocity of the crystal then produces a Coriolis force – an apparent force proportional to angular rate which is only present in the sensor coordinate system [17]. This force is directly proportional to the mass and angular velocity of the sensing element. Thus, the displacement of the sensing element is then proportional to the angular rate. In MEMS gyroscopes, a capacitive element is employed to track the Coriolis displacement of the sensing element. A schematic of an angular rate gyroscope is found in Fig. 11.

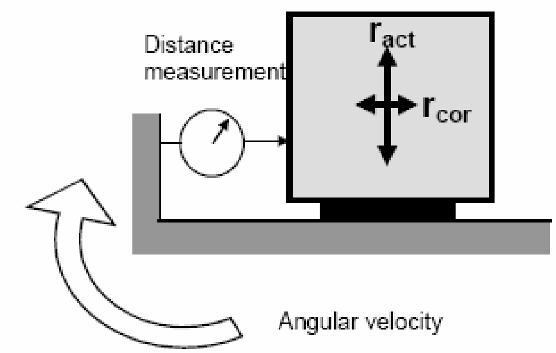

Fig. 11: Schematic diagram of gyroscope operating principle. The Coriolis force (r_{cor}) is in a rotating reference plane and remains perpendicular to the actuation force (r_{act}). Taken from [17].

3.2.4 Selection of Appropriate Gyroscope

Like the accelerometer, the gyroscope was subject to basic design constraints, including size and cost. Since the system was designed for motion capture in the sagittal plane, a single axis gyroscope is satisfactory. Empirical data taken from the literature suggests the natural angular rate of human limbs during gait to be about 90-120 degree per second [16]. The final design constraint on the gyroscope is the supply voltage. Vibration rate sensors require input voltages (12-16 V) larger than those supplied by standard data acquisition systems. Since the sensor is intended for use with standard systems, a sensor with integrated circuitry to amplify the input voltage is required.

The ADXRS300 from Analog Devices was chosen because it met all design criteria. The ADXRS300 is a +/- 300 deg/sec single axis gyroscope contained with required integrated circuitry on a single chip. The chip contains circuitry with a charge pump to amplify the input voltage from 5 V to the required 14-16 V. The output of the differential displacement capacitor is sent through demodulation stages included in circuitry on the chip, resulting in an analog voltage output proportional to angular rate. The ADXRS300EB evaluation board was also purchased. This board provides the necessary capacitors and resistors to set the bandwidth and charge pump for the device. As delivered, the ADXRS300 is set to a bandwidth of 40 Hz. A schematic of the ADXRS300 can be seen below in Fig. 12

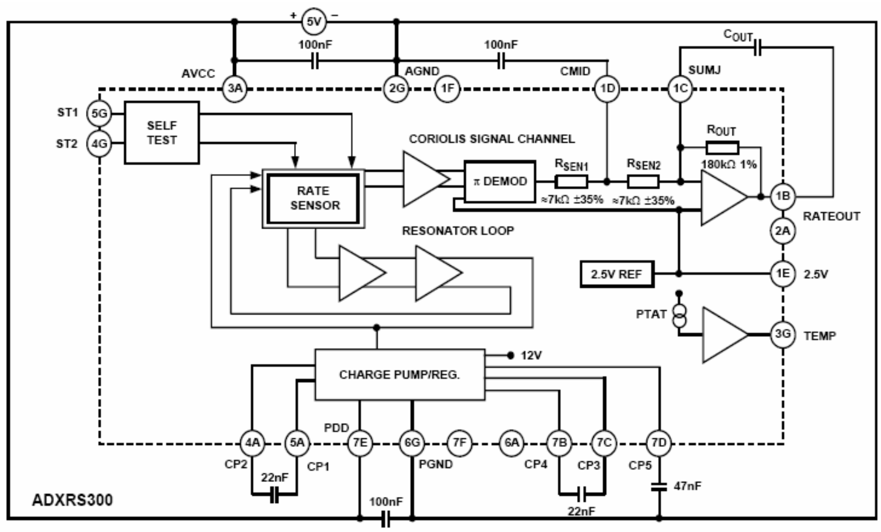

Fig. 12: Functional Block Diagram of the ADXRS300. Resistor R_{sen1} and Capacitor CMID are used to low pass filter signals at 400 Hz after demodulation but prior to amplification. Bandwidth is controlled by capacitor C_{out}. Taken from [21].

CHAPTER 4

Design of a Representative Mounting System

4.1 Base Bar Design

After the appropriate sensors were selected for planar motion capture, the mounting methods were designed. The mounting methods for the system were based directly on the successful design of Nene [14]. Rather than start with sensors directly attached to prosthetic leg components, a small aluminum bar was used to represent the actual prosthetic device. The bar provides an extremely cheap base for the system, while maintaining the benefits of rigid sensor attachment. The sagittal dimensions of the bar were chosen based on a rough estimate of anthropometric data for a natural leg combined with thigh and shank dimensions from prosthetic devices. Since all sensors to be mounted were extremely light, the thickness of the bar was constrained by patient comfort and material availability. The base bar can be seen in Fig. 13.

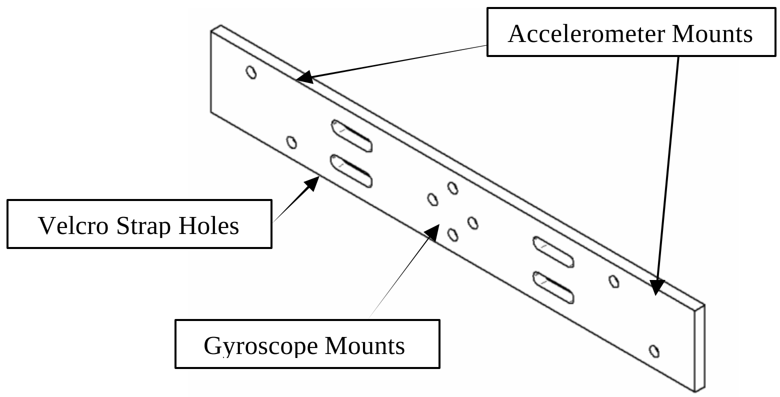

Fig. 13: Base bar for measurement system. Dimensions are 12.0 in x 1.75 in x 0.25 in.

The three (3) hole patterns in the base bar serve as attachment points for mounting plates. The peripheral sets are designed to hold the plates for the accelerometers, while the center set secures the gyroscope. The slots between the mounting patterns are for attachment of the bar to the leg via a Velcro strap.

4.2 Sensor Plate Design

The ADXL210 evaluation boards came with pre-fabricated mounted holes 0.88 inches apart. These holes were used to mount the accelerometer evaluation boards to aluminum plates, which were then attached to the base bar. Each plate was fabricated with 5 sets of holes to allow variation in the orientation of accelerometers with respect to the base (see Fig. 14). The plates were then attached to the periphery of the base bar by two screws, as seen in Fig. 15.

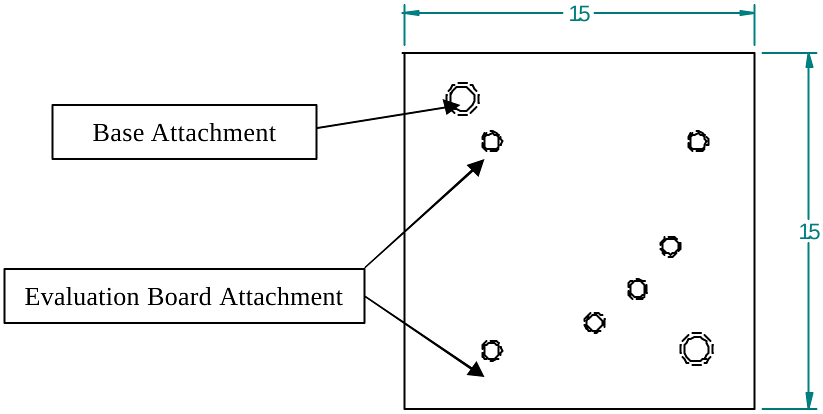

Fig. 14: Plate for attachment of accelerometer to base bar. Dimensions are 1.5 in x 1.5 in x 0.125 in. Holes of evaluation board attachment are spaced 0.88 in apart at angles of 0, 30, 45, 60, and 90 degrees.

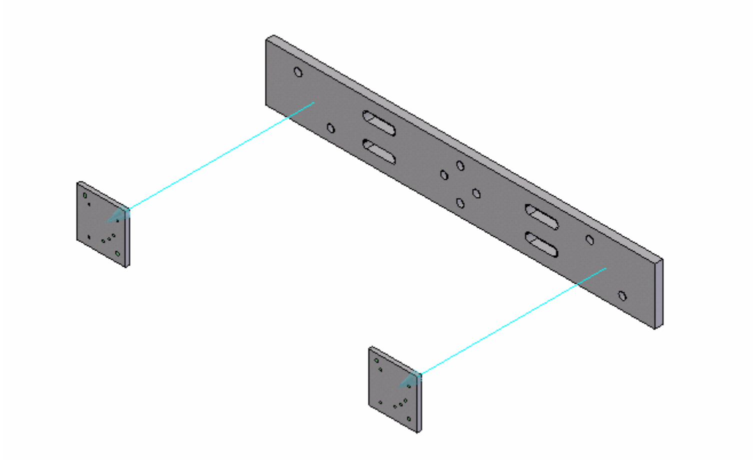

Fig. 15: Exploded view of base bar and accelerometer plates.

The ADXRS300 evaluation board was attached to a prototyping board in order to accommodate the required connectors. Only 7 of the 20 pins are necessary for the motion capture application. The prototyping board was then mounted directly to the base bar.

A photograph of the completed system, with sensors mounted, can be seen in Fig. 16.

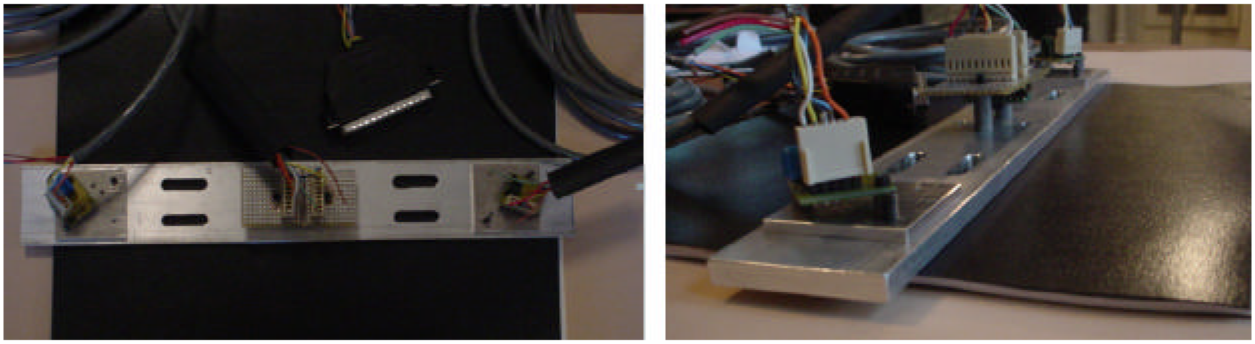

Fig. 16: Top and side view of assembled measurement system. ADXL210 accelerometers are mounted on the outside with the ADXRS300 mounted in the center.

CHAPTER 5
System Realization

5.1 Collected Signals and Proposed Algorithm

In a similar fashion to the work performed by Nene [14], a two dimensional, planar model (representing the sagittal plane) is formulated to obtain kinematic values from the sensor signals. The obtained sensor signals are as follows: $a_1^r, a_1^t, a_2^r, a_2^t, w$. The subscript on each acceleration signal represents the corresponding accelerometer position. The superscript denotes the direction of the acceleration signal vector- radial or tangential (See Fig. 17 below). The segment (bar) is assumed to be rigid during all calculations.

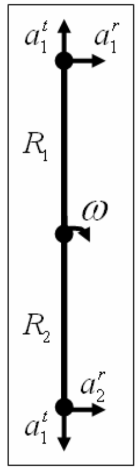

Fig. 17: Free body diagram of base bar with sensor signals represented as indicated above. R1 and R2 indicate the constant distance between the accelerometers and gyroscope.

Unlike Nene, the gyroscope placed at the center of mass of the segment (bar) is used directly to obtain the angle of the centroid. The calculation performed is:

$$q = \int_t w dt + q_{off} \qquad (4.1)$$

The offset can be found by examination of two perpendicular accelerations at the same point on the rigid bar [12]:

$$q_{off} = \arctan(\frac{a_1^t}{a_1^r}) \qquad (4.2)$$

The next step is to numerically integrate[1] the tangential acceleration, radial acceleration, and angular rate signals with respect to time. Then, the four (4) integrated accelerometer signals are integrated again to obtain position in the body coordinate system. The data is then rotated from the body coordinate system to the global coordinate system using the known bar geometry with the body coordinate angle form equation 4.1 in a homogenous transformation matrix.

[1] This and all other numerical integrations are performed using the cumulative trapezoidal method in MATLAB software.

5.2 Data Collection Hardware

Data collection will performed using a dSPACE data acquisition and control system. Real time data collection is made possible through the use of a DS1103 PPC controller board in combination with dSPACE and MATLAB software. The DS1103 PPC is a single board system housed in an expansion box external to the host PC. Communication between the controller board and the PC is accomplished using ISA-bus extension.

With the design of the hardware and software of the measurement system complete, the system was assembled and tested for initial competence. The ADXL210 evaluation boards contain 5 pin outs for connection to a data acquisition system (DAQ). A schematic of the pin configuration as well as their function can be seen in Fig. 18 below.

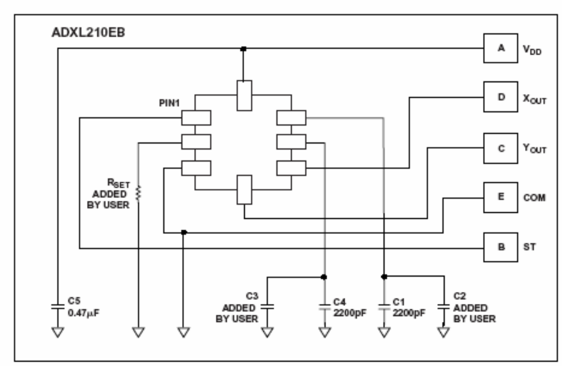

Table II. ADXL210EB Pinout Description

Pin Reference	Function
E	Ground
B	Self-Test Input
D	X-Axis Duty Cycle Out
C	Y-Axis Duty Cycle Out
A	+V Supply (3 V dc to 5.25 V dc)

Fig. 18: Schematic of ADXL210EB and Pinout Descriptions. Taken from [20].

The value used for R_{set} was 250 kO, giving a PWM period of about 2 ms. A capacitance of 0.1 μF was used for both C_1 and C_2, giving a nominal 50 Hz bandwidth with an RMS noise floor of 4.2 mg in both the X and Y directions. The two five (5) pin connectors from the accelerometers were wired into the Slave DSP PWM Signal Measurement port on the dSPACE breakout box via a Slave I/O Sub-D connector.

The ADXRS300 evaluation board contains integrated circuitry which demodulates the digital signal from the differential displacement capacitor on the actual gyroscope sensor chip. Therefore, the gyroscope signal was connected to the dSPACE system through a simple analog to digital port using a standard BNC connector.

5.3 System Implementation

Once the three (3) sensors were wired and connected to the dSPACE system they are fastened to the base bar. The experiment is then run by attaching the base bar to the limb of a subject using Velcro straps. A photograph of the system as it would be implemented in an experiment can be seen below in Fig. 19.

Fig. 19: Sensor system mounted in position for experimentation.

CHAPTER 6

Conclusion and Future work

6.1 Conclusion

This thesis presents an initial design for a portable, reliable, and cost effective measurement system for use in capturing artificial limb motion during an amputee's gait cycle. The measurement system developed is intended for use as a test-bed for establishing feasibility of gait monitoring using small, prosthetic mounted sensors. The system builds on previous work by formulating a new system based on positive results from literature. The body mounted sensor approach has been applied to motion tracking of human limbs [14], determination of dynamic loading on body joints [12], and recognition of events in the gait cycle [13]. The designed solution is intended for use as a gait monitoring system to aid in the clinical alignment process for lower limb prosthetic devices.

The measurement system uses two (2) accelerometers and a gyroscope to capture planar motion of body segments. Sensors were chosen based on their ability to track motion in low frequency applications. The mounting methods were designed to accurately represent the rigid nature of a prosthetic device.

6.2 Future Work

In this work, a design for tracking motion was proposed and realized. However, the feasibility of the suggested system has not been tested extensively. Immediate work is needed in order to evaluate this configuration with respect to accuracy, precision, and repeatability. In addition, further algorithm development and refinement is required in order to gain a true measure of device function. Data collection ability had been verified, but post-processing techniques have not been developed to the point of robustness. Evaluation of post-processing errors must be performed. These errors have historically been as influential as ones arising from hardware design [12-14].

While this system is based on previous work, research examining the quantification of prosthetic device alignment is still relatively young. A systematic approach to clinical alignment of lower limb prostheses is still distant. Development of systems similar to the proposed one offers an avenue for the evolution of prosthetic device alignment.

References

[1] Lannon, N. “Trans-tibial Alignment: Normal Bench Alignment.” *ISPO Ortholetter* 12. (2003) pp. 12-13.

[2] Blumentritt, S., et. al. “Effects of Sagittal Plane Prosthetic Alignment on Standing Trans-tibial Amputee Knee Loads.” *Prosthetics and Orthotics International.* 23.3 (1999) pp. 231-238.

[3] Sin, S.W., Chow, D.H.K, Cheng, J.C.Y. “Significance of non-level walking on transtibial prosthesis with particular reference to the effects of anterior-posterior alignment.” *Journal of Rehabilitation Research and Development.* 38.1 (2001) pp. 1-6.

[4] Zahedi, M.S., Spence, W.D., Solomonidis, S.E., Paul, J.P. “Alignment of lower-limb prostheses.” *Journal of Rehabilitation Research and Development.* 23.2 (1986) pp. 2-19.

[5] Reitman, J.S. et al. “Gait Analysis in Prosthetics: Opinions, Ideas and Conclusions.” *Prosthetics and Orthotics International.* 26.1 (2002) pp. 50-57.

[6] Sanders, J.E., et. al. “Effects of Changes in Cadence, Prosthetic Componentry, and Time on Interface Pressures and Shear Stresses of Three Trans-tibial Amputees.” *Journal of Clinical Biomechanics.* 15.9 (2000) pp. 684-694.

[7] McMahon, T.A. “Mechanics of Locomotion.” *The International Journal of Robotics Research.* 3.2 (1984) pp. 4-28.

[8] Middle East Technical University Department of Mechanical Engineering. Biomechanics Research Laboratory. <http://www.me.metu.edu.tr/biomechanics/Motion_Analysis/>

[9] Simon, S.R. “Quantification of human motion: gait analysis – benefits and limitations to its application to clinical problems.” *Journal of Biomechanics*. 37. (2004) pp. 1869-1880.

[10] Hailey, D., Tomie, J.A. “An assessment of gait analysis in the rehabilitation of children with walking difficulties. *Disability and Rehabilitation.* 22.6 (2000) pp. 275-280.

[11] Ko, W.H. “Solid-state physical transducers for biomedical research. *IEEE Transactions Biomedical Engineering.* 33.2 (1986) pp. 153-162.

[12] van den Bogert, A.J, Read, L., Nigg, B.M. “A method for inverse dynamic analysis using accelerometry.” *Journal of Biomechanics*. 29.7 (1996) pp.949-954.

[13] Tong, K., Granat, M.H. “A practical gait analysis system using gyroscopes.” *Medical Engineering and Physics.* 21.2 (1999) pp.87-94.

[14] Nene, A., Mayagoita, R., Vetlink, P. “Assessment of rectus femoris function during initial swing phase.” *Gait and Posture*. 9. (1999) pp.1-9.

[15] Jacobson, J.S., Kushekin, A.P. “An anthropomorphic highly functional artificial foot.” *Gait and Posture*. 5.1 (1997) pp. 77.

[16] Gowtizke, B.A., Milner, M. *Scientific bases of human movement*. 3rd ed. Baltimore: Williams & Wilkins. 1988. Ch. 2.

[17] Luinge, H.J. “Inertial Sensing of Movement.” Doctoral Thesis. Twente University Press. Enschede, Netherlands. (2002) pp.9-13.

[18] Lotters, J.C.O., Vetlink, P.H., Bergveld, P. “A sensitive differential capacitance to voltage converter for sensor applications.” *Instrumentation and Measurement*. 48.1 (1999) pp. 89-96.

[19] Kitchin, C. “Using Accelerometers in Low *g* Applications.” *Application Note AN-374*. Analog Devices. Norwood, Mass. <www.analog.com>

[20] Analog Devices ADXL210 Data Sheet Rev.0. <www.analog.com>

[21] Analog Devices ADXRS300 Data Sheet Rev.B. <www.analog.com>